INFLUENCE OF CONNECTOR DESIGN ON FRACTURE RESISTANCE OF TWO COMMERCIALLY AVAILABLE ALL CERAMIC FIXED PARTIAL DENTURES USING CAD/CAM

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<u>CERTIFICATE</u>

This is to certify that this dissertation titled "**INFLUENCE OF CONNECTOR DESIGN ON FRACTURE RESISTANCE OF TWO COMMERCIALLY AVAILABLE ALL CERAMIC FIXED PARTIAL DENTURES USING CAD/CAM"** is a bonafide record of work done by **Dr. C. KUMARAN** under my guidance and to my satisfaction during his postgraduate study period between 2013-2016.

This dissertation is submitted to **THE TAMILNADU Dr. M.G.R. MEDICAL UNIVERSITY**, in partial fulfillment for the award of the degree of **Master of Dental Surgery** in **Prosthodontics and Crown and Bridge, Branch I**. It has not been submitted (partial or full) for the award of any other degree or diploma.

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"No one who achieves success does so without acknowledging the help of others. The wise and confident acknowledge this help with gratitude."

- Alfred North Whitehead

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"Great teachers empathize with kids, respect them, and believe that each one has something special that can be built upon."

~ Ann Lieberman

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CONTENTS

LIST OF ABBREVIATIONS

Al_2O_3	Aluminium Oxide
CAD/CAM	Computer Aided Design / Computer Aided Manufacturing (or Milling)
CEREC	Chairside Economical Restoration of Esthetic Ceramics, or CEramic REConstruction
CNC	Computerized Numerical Control
FDP	Fixed Dental Prosthesis
FEA	Finite Element Analysis
FPD	Fixed Partial Denture
GE	Gingival Embrasure
GIC	Glass Ionomer Cement
OE	Occlusal Embrasure
PFM	Porcelain Fused to Metal
RoC	Radius of Curvature
STL	StereoLithography
Y-TZP – ZrO2	Yttrium Tetragonal Zirconia Polycrystal
	Zirconia Dioxide
ZnPO4	Zinc Phosphate

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INTRODUCTION

The rising interest in esthetic dentistry over the past decade has led to new materials and techniques to be developed in the quest for ultimate esthetic material[20]. This in turn has increased the demand for metal free restorations in the anterior as well as posterior region. Because of their esthetics and biocompatibility, many patients prefer All-ceramic crowns to metal-ceramic crowns. Nowadays, strong ceramic core materials have been developed to support the weaker veneering ceramic materials, particularly for the use of All-ceramic restorations in the posterior region.[20]

McLean, in 1967 introduced the idea of fabricating a high-alumina ceramic for the fabrication of fixed partial denture (FPD) pontic structures. In 1982, he introduced the platinum-bonded alumina FPD to reduce the problem of fracture through the connector area while eliminating the traditional cast metal framework[43]. However, this restorative option demonstrated a high rate of failure at the connector sites. Since then, developments in dental ceramics have led to the introduction of new high-strength ceramic core materials for All-ceramic FPDs.[43]

All-ceramic crowns provide a more desirable approach compared with metal–ceramic combinations for the restoration of natural dentition due to their superior aesthetics. The advantages of an All-ceramic FPD are the improved esthetics and lower allergenic potential of the ceramic materials used as substructure. However, ceramics are brittle and susceptible to failure beyond a critical stress, due to surface flaws. Current All-ceramic materials offer an accepted level of fracture resistance, fit and esthetics. High fracture resistance recommends it to be a material to support fixed partial denture (FPD) in a stress bearing area with clinical success.

The clinical fracture resistance of FPDs is related to the size, shape, and position of the connectors and to the span of the pontic. The basis for the proper design of the connectors and the pontic is the law of beams: deflection of a beam increases as the cube of its length, it is inversely proportional to its width, and it is inversely proportional to the cube of its height[43].Of all the structural factors, the connector areas are the most influential in failure. In All ceramic fixed partial dentures the connector area is a common fracture location. Failure rate is relatively high in three unit All-ceramic bridges around the sharp connector area[33]. The FPD shape is not uniform clinically, but is a complex combination of multiple convexities and concavities that depend on the geometry and alignment of the teeth. The high incidence of connector fracture in posterior areas may be attributed to lower inter-occlusal space and short clinical crowns. Furthermore, the connector area is usually narrowly constricted for biological or esthetic reasons, which typically concentrates stresses relative to the average stress levels in other areas of the prosthesis[35].When occlusal forces are applied directly through the long axis of an All-ceramic FPD at the midspan (pontic), compressive stresses will develop at the occlusal aspect of the connector at the marginal ridge, and tensile stresses will develop at the gingival surface of the connector[36]. These tensile stresses contribute to the propagation of micro cracks located at the gingival surface of the connector through the core material in an occlusal direction, and may eventually result in fracture. The survival time of three-unit fixed partial dentures may be improved by altering the connector design in regions of maximum tension.

Previously, it had been hypothesized that fracture initiation sites in dental ceramics could be controlled by changing the ceramic thickness. However, it is now believed that ceramic thickness plays a secondary role in fracture initiation, and that critical flaws in regions of reduced thickness are generally more important.

All ceramic prosthesis has become popular and can be fabricated by both traditional laboratory methods and CAD/CAM. The traditional methods were time-consuming, and technique sensitive. CAD/CAM may be a good alternative as it reduce the fabrication time of high strength ceramics by up to 90%. The ceramic blocks used for CAD/CAM are more homogenous with minimal flaws compared with laboratory processed dental ceramics.

The purpose of this study was to evaluate the fracture resistance of two commercially available CAD/CAM manufactured All ceramic FPD restorations by varying the radii of curvature of gingival embrasure.

AIM & OBJECTIVES

Aim:

To evaluate and compare the fracture resistance of two commercially available (CeramillZi and Lava plus) three unit All ceramic bridge copings in 0.45 mm and 0.9 mm radii of curvature at the gingival embrasure.

Objectives:

The purpose of this study is to compare the fracture resistance of tooth supported 3 unit All ceramic fixed partial denture copings with two different connector designs (radii of curvature at gingival embrasure).

1. To determine the fracture resistance of CeramillZi (AmannGirbach) three unit All ceramic bridge copings in 0.45 mm radius of curvature at the gingival embrasure.
2. To determine the fracture resistance of Lava Plus (3M ESPE) three unit All ceramic bridge copings in 0.45 mm radius of curvature at the gingival embrasure.
3. To determine the fracture resistance of CeramillZi (AmannGirbach) three unit All ceramic bridge copings in 0.9 mm radius of curvature at the gingival embrasure.
4. To determine the fracture resistance of Lava Plus (3M ESPE) three unit All ceramic bridge copings in 0.9 mm radius of curvature at the gingival embrasure.
5. To compare the fracture resistance of CeramillZi (AmannGirbach) and Lava Plus (3M ESPE) three unit All ceramic bridge copings in 0.45 mm and 0.9 mm radii of curvature at the gingival embrasure.

REVIEW OF LITERATURE

- **Chen HY, Hickel R, Setcos JC, Kunzelmann KH** [7] (1999) conducted a study to determine the fracture strength of various all-ceramic crowns, with and without prior cyclic loading. Standardized molar crowns were fabricated with a CAD-CAM machine (Cerec 2), software with machinable ceramic materials (Vita Mark II and ProCAD), and also conventional heat-pressed IPS Empress crowns fabricated at two dental laboratories. They concluded that Cerec ProCAD crowns demonstrated significantly greater strength than the Vita Mark II crowns, better resistance to cyclic loading and lower failure probability than the laboratory-fabricated IPS Empress crowns.
- **Tinschert J, Natt G, Mautsch W, Augthun M, Spiekerman H**[59] (2001) conducted a study to determine the fracture resistance of three-unit fixed partial dentures (FPD) made of new core ceramics (IPS Empress 2, In-Ceram Alumina, In-Ceram Zirconia, and DC Zirkon). A base metal three-unit master FPD model with a maxillary premolar and molar abutment was made. FPDs were constructed with a uniform 0.8-mm-thick core ceramic and a porcelain veneer layer. All FPDs were cemented with ZnPO4 on the master model and loaded on a universal testing machine until failure. The authors concluded that the high fracture resistance evaluated for FPDs made of DC-Zirkon underscores the remarkable mechanical properties of high performance ceramic, which could be useful for highly loaded all-ceramic restorations, especially in the molar region.
- **Oh WS, AnusaviceKJ**[36] (2002) suggested that the fracture probability may be significantly reduced by using a connector with a curvature radius of approximately 0.9 mm. To reduce the stress concentration and to maintain a

constant connector height of 4.0 mm without altering the curvature at the gingival embrasure, they suggested a gingival embrasure curvature radius of 0.45 mm. This propagates the cracks from the gingival embrasure toward the occlusal loading on the pontic. The fracture origin was most commonly at the center of the gingival embrasure in the buccolingual dimension and was shifted slightly toward the abutment crown in the mesiodistal direction.

- **Luthy H, Filser F, Loeffel O, Schumacher M, Gauckler LJ, Hammerle CH**[30] (2005) conducted an in-vitro study, to compare the fracture strength of four-unit Y-TZP FPD cores designed with different connector diameters. A total of 40 four-unit FPD cores supported by end abutments and having two pontics were manufactured in Procera Zirconia. Five groups of FPD cores with connector dimensions of 2·0, 2·5, 3·0, 3·5 and 4·0 mm were produced. Fracture strength was significantly higher for each increase in connector diameter except for the 2·0-mm and 2·5-mm diameters where all fractures occurred during preload. All FPD cores fractured in the connector area. Within the limitations of thisstudy, a minimum diameter of 4·0 mm is recommended for All-ceramic zirconia-based FPDs with long spans or replacing molars. Clinical studies are, however, needed to determine adequate connector dimensions.
- **Sailer I, Feher A, Filser F, Lüthy H, Gauckler LJ, Schärer P et al**[49] (2006) conducted a study to determine the success rate of 3 to 5 unit posterior fixed partial dentures (FPDs) with zirconia frameworks after 3 years of functioning. Clinical and radiographic examinations were performed at baseline, 12, 24, and 36 months after cementation. Statistical analysis was performed by

descriptive statistics and the Kaplan-Meier survival analysis. Comparisons of probing depth, plaque index, and bleeding on probing between test (abutment) and control (contralateral) teeth was done with the McNemar test. There were no significant differences regarding the probing depth in test and control teeth. It was concluded that zirconia frameworks demonstrated sufficient stability for replacement of posterior teeth. However, the high rates of technical problems should be reduced by further developments of the prototype processing technology.

- **Motta AB, Pereira LC, da Cunha ARCC[33]** (2007) conducted a 2D finite element study to compare the stress distribution on 3-unit All-ceramic and metal-ceramic fixed partial dentures (FPDs) and to identify the areas of major risk of failure. Three FPD models were designed: (1) metal-ceramic FPD; (2) All-ceramic FPD with the veneering porcelain on the occlusal and cervical surface of the abutment tooth; (3) All-ceramic FPD with the veneering porcelain only on the occlusal surface. A 100 N load was applied in an area of 0.5 mm^2 on the working cusps, following these simulations: (1) on the abutment teeth and the pontic; (2) only on the abutment teeth; and (3) only on the pontic. When the load was applied on the pontic, the highest stress values appeared on the connector areas between the abutments and pontic. In conclusion, the best stress values and distribution were found for the All-ceramic FPD with the veneering porcelain only on the occlusal surface. However, in under clinical conditions, fatigue conditions and restoration defects must be considered.

- **Esquivel-Upshaw JF, Young H, Jones J, Yang M, Anusavice KJ** [17] (2008) conducted a study with the hypothesis that 3-unit fixed partial dentures (FPDs) made from a moderately high-strength core ceramic will adequately resist fracture in posterior regions if fabricated with a minimal connector size of 4 mm. Connector heights and widths were measured for each FPD. All FPDs were examined by 2 independent clinicians, and rankings for each criterion were made from 1 to 4 (4 = excellent; 1 = unacceptable). From a fracture resistance perspective, 4 of the 30 ceramic FPDs fractured within the 4-year evaluation period, representing an 86.7% success rate. Another FPD was replaced because of a caries lesion on 1 abutment tooth away from the margin. One FPD fracture was associated with the subject having the greatest occlusal force (1,031 N). The other 2 fractures were associated with FPDs that exhibited connector heights of less than 3 mm. All criteria were ranked good to excellent during the 4-year period for the remaining FPDs. They concluded that fractured FPDs were associated with a connector height of less than 4 mm.
- **Bahat Z, Mahmood DJ, Vult von Steyern P** [4] (2009) conducted a study to investigate how different radii of curvature in the embrasure of the connector area and different connector dimensions could affect the fracture resistance of 3-unit All-ceramic fixed partial dentures (FPDs) made of Y-TZP. Forty-eight FPDs in 6 groups of 8 FPDs with different connector design were produced in Procera Zirconia Bridge material. All the FPDs fractured in the connector area. All the crack propagation which led to fracture started at the gingival embrasure of the connector. Within the limitations of this in-vitro study, the recommended minimum dimension of an anterior 3-unit All-ceramic FPD of

Y-TZP is 3 mm in incisal-cervical direction and 2mm in buccal-lingual direction. By increasing the radius of the gingival embrasure from 0.6 to 0.9 mm, the fracture strength for a Y-TZP FPD with connector dimension 3 x 3 mm increased by 20%.

- **Inan O, Secilmis A, Eraslan O**[20] (2009) compared the fracture resistance of implant-supported All-ceramic fixed partial dentures, which have three different pontic designs. Thirty standardized 3-unit All-ceramic fixed partial dentures with biconvex, convex or concave pontic designs were fabricated using IPS e.max system. The fracture resistance values of All-ceramic fixed partial dentures designed with biconvex, convex or concave pontics was 349.71, 438.20 and 300.78 N, respectively. There were no statistically significant differences between the fracture resistances of the groups ($p>0.05$), except for convex and concave groups ($p<0.05$ and $p=0.009$, respectively).
- **Plengsombut K, Brewer JD, Monaco EA Jr, Davis EL**[39] (2009) conducted a study to determine the effect of 2 connector designs on the fracture resistance of core materials used for All-ceramic fixed dental prostheses (FDPs). Three materials were tested: (1) heat-pressed lithium disilicate glass ceramic (IPS e.max Press), (2) milled lithium disilicate glass ceramic (IPS e.max CAD), and (3) milled yttrium-stabilized tetragonal zirconia polycrystals (Y-TZP) (IPS e.max ZirCAD). Specimens were made into 30 x 4 x 4-mm bars to represent 3-unit FDPs. Two connector designs, round (0.60 ±0.01-mm radius of curvature) and sharp (0.06 ±0.001- mm radius of curvature), with a 3.00 ±0.05-mm cross-section for each connector, were studied (n=5). The authors concluded that the fracture resistance of ceramic core materials is affected by

fabrication technique and connector design, connector design affected fracture resistance of the milled ceramic, but not the pressed ceramic, and a round connector design is preferred for maximizing fracture resistance of All-ceramic FDP framework materials for CAD/CAM.

- **Quinn GD, Studart AR, Hebert C, VerHoef JR, Arola D**[41] (2010) conducted a study to estimate the maximum stress concentration posed by the connector geometry and to provide adjusted estimates of the minimum connector diameter that is required for achieving 20 years of function. Results showed that the magnitude of stress concentration estimated for clinically relevant connector geometries ranges from 2 to 3. The author concluded that adjusted estimates for the minimum connector diameter required to achieve 20 years of clinical function was found to be around 4 mm.
- **Senyilmaz DP, Canay S, Heydecke G, Strub JR**[52] (2010) conducted a study to evaluate the fracture resistance and the survival rate of various All-ceramic crowns in-vitro after thermo mechanical fatigue loading in comparison to porcelain-fused-to-metal posterior crowns. Sixteen crowns for human mandibular first molars were made of each of the following: Cercon, IPS-Empress 2 In-Ceram Zirconia, Procera AllZircon and porcelain-fused-to-metal. All samples were tested for the maximum fracture resistance. The survival rates after 1-2 million cycles in the artificial mouth were 100% in all the tested crown systems. The chewing simulation and thermocycling did not significantly decrease the fracture strength of the ceramic crowns ($P>0.005$). The median fracture load of Cercon, Procera AllZircon, In-Ceram Zirconia and PFM was significantly higher than IPS-Empress 2 both for loaded and

non-loaded groups ($P<0.005$) while the difference between Cercon, Procera AllZircon, In-Ceram Zirconia and PFM was not significant ($P>0.005$). All-ceramic systems showed fracture load values similar to those of porcelain-fused-to-metal molar crowns and therefore may be considered for use in clinical studies.

- **Onodera K, Sato T, Nomoto S, Miho O, Yotsuya M**[38] (2011) conducted a study to determine the relationship between cross sectional design and fracture load using a static load bearing test in yttria-stabilized tetragonal zirconia polycrystal (Y-TZP) ceramic frameworks on a molar fixed partial denture. The cross sectional area of the connector was 9, 7, 5 mm^2. In terms of shape, the cross section was either circular or oval. In terms of cross sectional area, there was a statistically significant difference in the fracture load between 9, 7, 5 mm^2. No significant difference in fracture load was observed between any 2 shapes of the connector. Fracture occurred at the distal connector in 82.2% of all frameworks on average. Fracture load decreased as cross sectional area of the connector became smaller.
- **Dittmer MP, Kohorst P, Borchers L, Schwestka-Polly R, Stiesch M**[13] (2011) determined the stress distribution in four-unit fixed partial dentures (FPDs) made of yttria-stabilized polycrystalline tetragonal zirconia (Y-TZP), under an occlusal load. A three-dimensional finite element model was constructed and a stress analysis performed with a force of 1630 N applied at the centre of the middle connector area. The maximum tensile stress in the area of the middle connector amounted to 633 MPa. It increased with the load being applied from the oral towards the buccal side (648 MPa) and decreased with the load being applied from the buccal towards the oral side (570 MPa).

- **Rezaei SMM, Heidarifar H, Arezodar FF, Azary A, Mokhtarykhoee S**[46] (2011) conducted a study to determine the effect of buccolingual increase of the connector width on the stress distribution in posterior fixed partial dentures made of IPS Empress 2. The buccolingual connector width varied from 3.0 to 5.0 mm. Bridges were vertically loaded with 600 N at one point on the central fossa of the pontic, at 12 points along the cusp-fossa contact (50 N each), or at eight points along the cusp-marginal ridge contact (75 N each). Alternatively, a load of 225 N was applied at a 45º angle from the lingual side. The results showed that stress concentrations were observed within or near the connectors. The von Mises stress decreased by increasing connector width, regardless of whether the loading was applied vertically or at an angle. They concluded that that increasing the connector width decreases the failure probability when a vertical or angled load is applied.
- **Kohorst P, Junghanns J, Dittmer MP, Borchers L, Stiesch M**[23] (2011) conducted a study to evaluate the influence of different processing routes on the fitting accuracy of four-unit zirconia fixed dental prostheses (FDPs) fabricated by computer-aided design/computer-aided manufacturing (CAD/CAM). Three groups of zirconia frameworks with ten specimens each were fabricated. Horizontal marginal discrepancy, vertical marginal discrepancy, absolute marginal discrepancy, and marginal gap were evaluated. Statistical analysis revealed that, with all measurements, the marginal accuracy of the zirconia FDPs was significantly influenced by the processing route used ($p < 0.05$). Within the limitations of this study, all restorations showed a clinically acceptable marginal accuracy; however, the results suggest

that the CAD/CAM systems are more precise than the CAM-only system for the manufacture of four-unit FDPs.

- **Sannino G, Pozzi A, Schiavetti R, Barlattani A**[51] (2012) conducted a study to investigate the influence of different loading conditions on the stress distribution in a 3-unit implant supported Y-TZP fixed partial denture (FPDs) by finite element analysis and fatigue analysis. 100 N and 300 N loads over 0.5 mm^2 areas with different angles (0^0, $15^0$and 35^0) and locations were applied on the prosthesis and the distribution of equivalent von Messes stress was investigated. Results showed that maximum stresses were found at the connector region of the framework when the intermediate element was loaded. No fracture fatigue occurred with 100N force. A 300 N force applied to pontic produced no fatigue problems but when applied to one or both the pillars generated fatigue problem. The authors concluded that Y-TZP showed a structural reliability as framework material for 3-unit posterior FPDs.
- **Salimi H, Mosharraf R, Savabi O**[50] (2012) conducted a study to evaluate the effect of different framework designs on fracture resistance of zirconium oxide posterior fixed partial denture (FPD). Thirty two posterior zirconia FPD cores were manufactured to replace a second premolar. The specimens were divided into 4 groups; 1 – 3 x 3 mm connector and standard design, 2 – 3x 3 mm connector and modified design, 3 – 4 x 4 mm connector and standard design, 4 – 4 x4 mm connector and modified design. Results showed that the mean fracture resistance of group with 4 x 4 mm connector was significantly higher than groups with 3 x 3 mm connector ($p<0.001$). The authors concluded that the fracture resistance of Zirconia posterior FPD was significantly affected by the connector size and not by the framework modification.

- **Kermanshah H, Bitaraf T, Geramy A**[22] (2012) conducted a study to determine the effect of trenched zirconia bar on the von Mises stress distribution of IPS –Empress II core ceramics. The model was reinforced with zirconia bar (ZB), zirconia bar with vertical trench (VZB) and zirconia bar with horizontal trench (HZB) (cross sections of these bars were circular). The model without zirconia bar was designed as the control. The bridges were loaded by 200 N and 500 N on the occlusal surface at the middle of the pontic component and von Mises stresses were evaluated along a defined path. The authors concluded that the embedded trenched zirconia bar could reinforce IPS-Empress II at the connector area which is a main failure region in all ceramic fixed partial dentures.
- **Ambre MJ, Aschan F, Vult von Steyern P**[3] (2013) conducted a study to investigate the fracture strength and fracture mode of yttria-stabilized tetragonal zirconia polycrystal (Y-TZP) posterior three-unit fixed dental prostheses (FDPs) with varying connector dimension and abutment core thickness. Groups with the same connector dimension showed no significant difference in fracture strength. All fractures of the specimens involved the connector. Within the limitations of this in vitro study, it can be concluded that the strength of an all-ceramic Y-TZP FDP beam depends more on the connector dimension than on the thickness of the abutment core.
- **Takuma Y, Nomoto S, Sato T, Sugihara N**[57] (2013) conducted a study to test static load bearing force on 4 unit Y-TZP All ceramic fixed partial denture (FPDs) with different cross sectional areas and forms to evaluate the influence of connector design on fracture load. Each of the central, mesial and distal connector was prepared with one of the two different cross sectional areas and

one of the three different forms (one circular and two oval forms) to give a total of 18 designs.The authors concluded that sufficient height needs to be maintained in the mesial/distal connector to secure a high fracture load in zirconia 4 units All ceramic FPDs. High fracture load may also be obtained by making the height of the central connector as great as possible. Also, extending the connector cross sectional area is effective in increasing fracture load.

- **Singh MS, Tuli AK**[54] (2014) conducted a study to investigate the effect of variable cross-sectional design of the connectors in stress distribution, in 3 different All ceramic core materials. Finite element analysis was conducted on 18 models using 3 differentceramic core materials i.e. IPS Empress II, In-Ceram Alumina and Zirconia polycrystals and 3 different connector designs i.e. circular, triangular and inverted T shape.: Maximum stress concentrations were observed in Empress 2 material and the leastwere seen with the Zirconia and intermediate in the In-Ceram Alumina fixed partial dentures. Also, of the three experimental connector designs, the maximum stress concentration was seen with circular cross sectional design while the least was with inverted T shape connector. The study concluded that maximum stress concentration in a three unit All ceramic fixed partial denture was affected by the modification of the cross sectional design of the connector.
- **El-Naga AAA, El-fallal AAE, Ibraheim SAAEF, Alaraby HA** [14] (2014) conducted an in-vitro study to investigate the effect of intaglio surface conditioning on the fracture strength of veneered Zirconia crowns compared to full-contour Zirconia crowns. The specimens were divided in to two groups (N=16) for each ceramic material. In the first group all prepared premolars

restored by fullcontour zirconia crowns (CAD/CAM) while in second group all teeth restored by veneered zirconia crowns. Each group was subdivided into two subgroups (N=8).Crowns of first subgroup were sandblasted with 50μm Al_2O_3 while the second subgroups were not treated by sandblast. The authors concluded that full-contour zirconia showed higher fracture resistance than veneered zirconia crowns. Surface treatment (sandblasting) has no effect on fracture strength mean.

- **Nassef TM, Khalil MF, Kader SH**[56] (2014) conducted a study to determine the influence of different connector designs on the fracture resistance and the stress distribution in posterior three units Zirconia fixed partial dentures (FPDs) of Incoris-TZI material. Twelve three units full contoured Zirconia fixed partial dentures were constructed by using CEREC 3 CAD/CAMsystem. The models are divided into two groups each of six according to the different radius of curvature of the occlusal embrasure (OE) and the gingival embrasure (GE) with the same connector height width ratio of 4:3. Group A*: Design (I)* OE 0.90mm GE 0.25mm.Group B: *Design (II)* OE 0.25mm GE 0.90mm. The results showed that the fracture initiation was at the gingival embrasure. Statistically significant difference between design (I) and design (II) zirconia FPDs when tested for fracture resistance. Design (II) showed higher fracture resistance than design (I) according to (OE-GE) the mean value of stress analysis was higher in the gingival embrasure than that of occlusal embrasure.
- **Correia ARM, Fernandes JCS, Campos JCR, Vaz MAP, Ramos NVM**[10] (2014) conducted a study to analyze the stress distribution on a cantilever-fixed partial denture (FPDs) after simulation of maximum mastication loads. A cantilever-fixed partial denture framework was designed in the CAD-CAM

system EverestKavo v2.0 using two materials, titanium and zirconium, with connectors of 5.28mm^2 and 9.05mm^{2}respectively.The authors concluded that titanium cantilever fixed partial denture frameworks with a 5.28mm^2 connector area cannot support maximum mastication loads; frameworks of this material require larger connectors with fillets introduced in the gingival embrasure. Zirconia, however, supports maximum bite forces in most situations with both molar and premolar design cantilevers. Precaution should be taken when dealingwith smaller connectors of 5.28mm^2.

- **Rismanchian M, Shafiei S, Nourbakhshian F, Davoudi A[47]** (2014) conducted a study to evaluate the flexural strength of two different zirconia frameworks for impact forces in implant supported fixed dental prostheses(FDPs). Two implant abutments with 3.8 mm and 4.5 mm platform were used as premolar and molar. They were mounted vertically in an acrylic resin block. A model with steel retainers and removable abutments was fabricated by milling machine; and 10 FDP frameworks were fabricated for each Biodenta and Cercon systems. All samples were thermo-cycled for 2000 times in 5-55˚C temperature and embedded in 37˚C artificial saliva for one week. The authors concluded that there was no significant difference between flexural strengths of both zirconia based framework systems; and both Biodenta and Cercon systems are capable to withstand biting force (even parafunctions) in posterior implant-supported bridges with no significant differences.
- **Olio M, Kvam K, Gjerdet NR[37]** (2014) conducted a study to compare three All ceramic systems (Alumina core, zirconia core, glass ceramic core) using a clinically relevant test method able to simulate clinical failure modes. Ten

incisor crowns of three types of All ceramic systems each were exposed to soft loading until fracture. The initiation and propagation of cracks in the crowns were compared with those of a reference group of crowns that failed during clinical use. All crowns fractured in a manner similar to fracture of the clinical reference crowns. The zirconia crowns fractured at statistically significantly higher loads than alumina and glass ceramic crows. Fracture initiation was in the core material, cervically in the approximal areas.

- **Murase T, Nomoto S, Sato T, Shinya A, Koshihara T, Yasuda H[34]** (2014) conducted a study to evaluate the effect of cross sectional area and morphology of the connector on its strength in yttrium tetragonal zirconia polycrystal (Y-TZP) All ceramic fixed partial dentures (FPDs) for the mandibular incisor region by fracture tests. Nine types of zirconia framework for 3 units FPD each differing in cross sectional area and morphology was prepared. Fracture load differed significantly among all groups according to cross sectional area and was also greater when the shape of the connector formed an isosceles triangle wide at the base and the connector had the same height and width. The authors concluded that connector design affected fracture load.
- **Lakshmi RD, Abraham A, Sekar V, Hariharan A[27]** (2015) conducted a study to analyze the stress distribution between monolithic Lithium-disilicate and monolithic Zirconia inlay retained Fixed Dental Prostheses by varying the connector dimensions using the 3D- Finite Element Analysis. Two models of three unit inlay retained Fixed Dental Prosthesis (FDPs) replacing the lower right first molar was fabricated, each with the connector dimensions of 3 mm x 3 mm and 4 mm x 4 mm. These groups which were then subjected to a vertical

load of 500N directed occlusally over a surface area of 5 mm^2.The results showed that by increasing the connector dimensions up to 4 mm x 4 mm, both the materials were capable of withstanding a force of upto 500 N, simulating the maximum posterior bite force. The authors concluded that monolithic Zirconia and Lithium-di-silicate can be used a posterior restorative material in All ceramic inlay retained Fixed Dental Prosthesis.

MATERIALS & METHODS

MATERIALS USED IN THE STUDY

The following materials were used for the study:

1. Custom fabricated stainless steel metal die simulating a sectional FPD model to replace missing left first molar. (Fig. 1)
2. Materials used for CAD/CAM generated restorations
    a. Yttria partially stabilized Zirconia blank (Y – TZP ZrO2: **CeramillZi,** AmannGirrbach AG, Austria) (Fig. 2)
    b. Yttria – stabilized tetragonal zirconia blank (Y – TZP **Lava Plus**,3M ESPE, Germany) (Fig. 3)
    c. Contrast spray (Ivoclar, Germany) (Fig. 4)
3. Adhesive resin cement (Bifix, VOCO, America) (Fig.5)

EQUIPMENTS USED IN THE STUDY

1. Equipments used for metal die fabrication
    a. Makino Vertical Machining Centre-Model S33 (Makino Milling Machine Co Ltd, Singapore) (Fig. 6)
    b. Solid Work 2006, CAM software (Concord, Massachusetts, USA)

2. Equipments used for CAD/CAM generated restorations
    A. CeramillZi (Y – TZP ZrO2:AmannGirrbach AG, Austria)
        i. CeramillMap 300scanner (Fig. 7)
        ii. CeramillMind software
        iii. CeramillMotion 2 (4X) milling unit (Fig. 8)
        iv. CeramillTherm Sintering unit (Fig. 9)

B. Lava Plus(Y – TZP, 3M ESPE, Germany)

i. Exocad Software

ii. LAVA CNC 240 milling unit (Fig. 10)

iii. Nabertherm Sintering unit (Fig. 11)

3. Equipments used for testing fracture resistance:

Universal testing machine (Zwick/Roell, Leominster, UK) (Fig. 12)

METHODOLOGY

The methodology of the study is shown in figure and it was divided into the following stages:

I. Fabrication of metal dies

II. Fabrication of ceramic copings

a. Fabrication of CeramillZi CAD/CAM copings

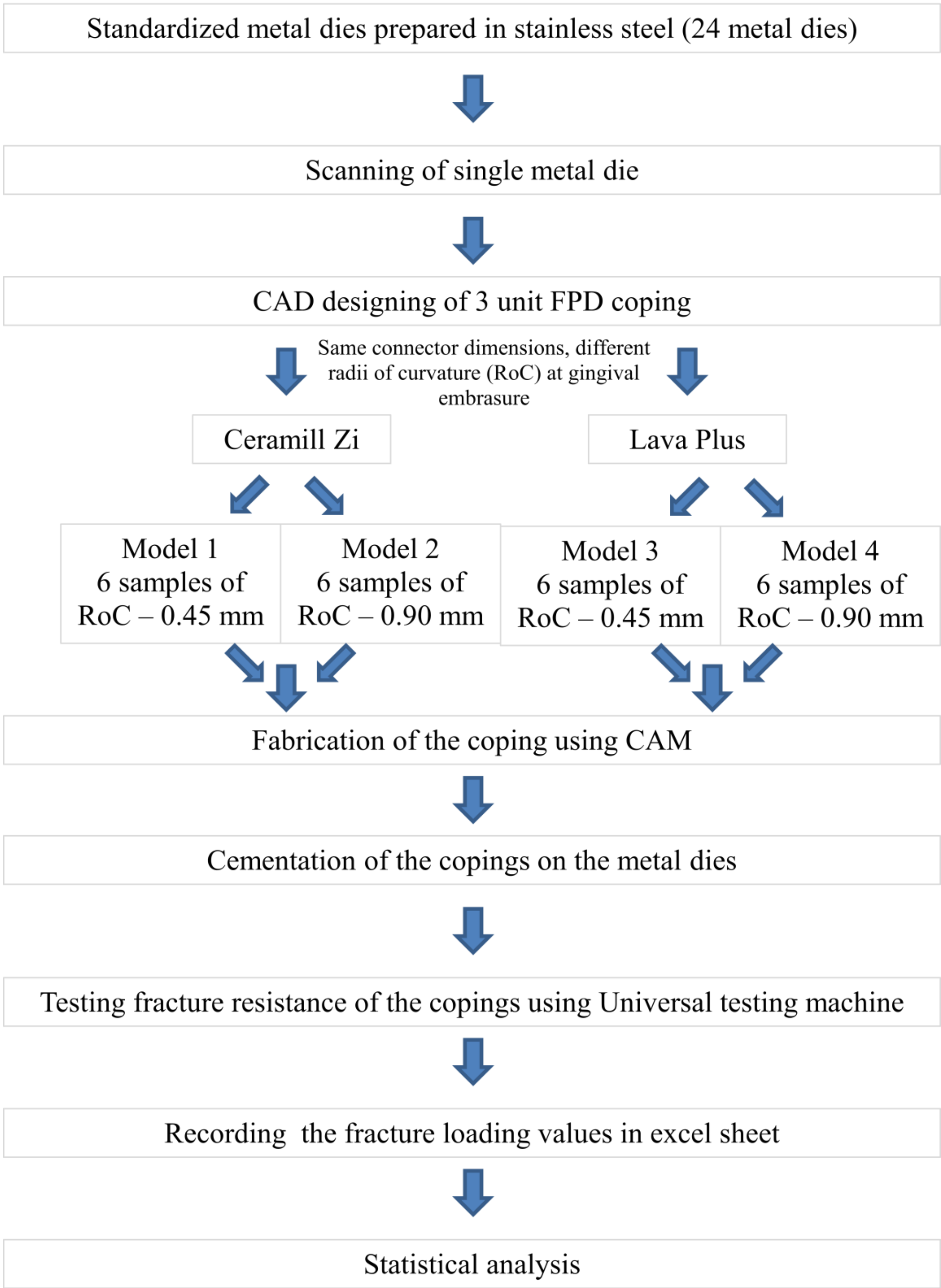
b. Fabrication of Lava Plus CAD/CAM copings

III. Cementation of the samples

IV. Testing samples under Universal Testing Machine

V. Statistical analysis

Fig13 - Methodology of the study



I. FABRICATION OF METAL DIE

24 metal dies were designed and milled to simulate a sectional 3 unit FPD model replacing mandibular left first molar with second premolar and second molar as the abutment teeth (Fig. 14) in the Centre for Advanced Tooling and Precision Dies, (PSG College of Engineering, Coimbatore). The metal dies were designed using Solid Work, CAM software. Stainless steel metal rods were milled in Makino Vertical Machining Centre-Model S33 (Makino Milling Machine Co Ltd, Singapore).

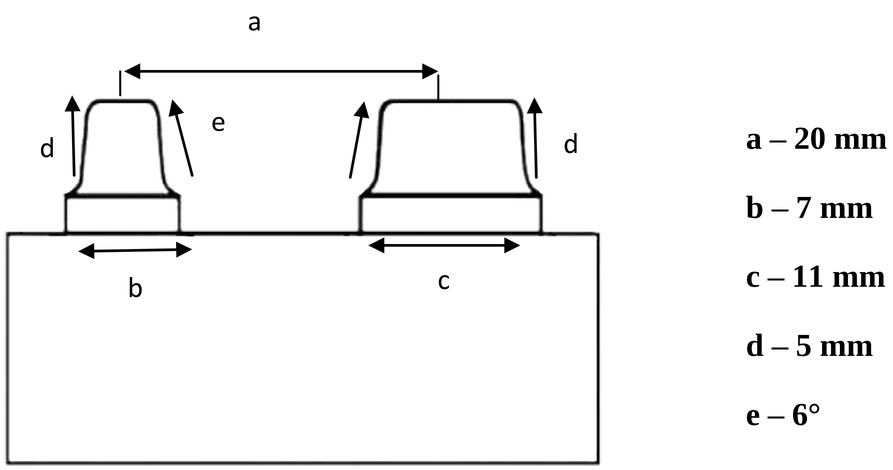


Fig. 15 - Schematic diagram of the metal die

The dimensions of metal dies were as follows[57]: (Fig. 15)

a. The distance between abutments was 20.0 mm

b. Diameter of the abutment was 7.0 mm corresponding to the second premolar

c. Diameter of the abutment was 11.0 mm corresponding to the second molar

d. The abutment had a height of 5.0 mm

e. The axial surface had a taper of 6°

f. The finish line was designed as a shoulder of 1.2 mm

II. PREPARATION OF CERAMIC COPINGS

12 three unit zirconia copings were constructed from CeramillZi and Lava Plus blanks by using Ceramill CAD CAM System (Vitallium dental lab, Chennai) and Lava CAD CAM System (Radiance Dental Fusion Pvt. Ltd, Bengaluru) each respectively. A total of 24 three unit bridge coping samples were divided into four groups, each group containing six samples according to the design of the fixed partial denture connector. The designed coping was[38] (Fig. 16)

a. 0.5 mm in thickness with flat occlusal surface
b. Mesio distal width of pontic was 8.0 mm
c. Bucco lingual width of pontic was 10.0 mm

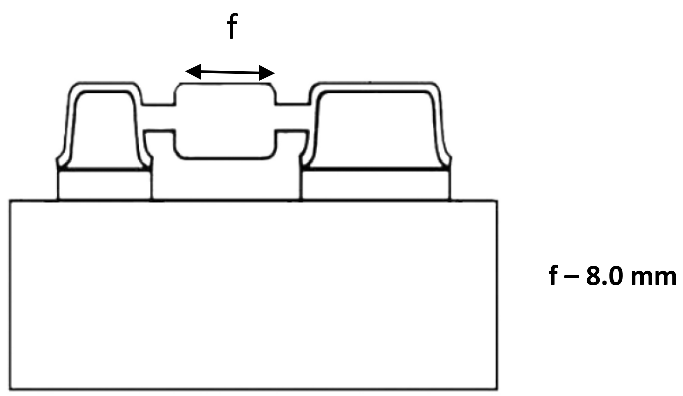


Fig 16 - Schematic diagram of die with coping

The connector dimensions for both the groups were standardized as:

a.	Cross section	Circular
b.	Height x Width x Length	4 x 4 x 2 mm
c.	Occlusal Embrasure (RoC)	0.25 mm

The two connector designs included in the study were differing in the radius of curvature (RoC) of the gingival embrasure (GE) [0.45 mm and 0.90 mm].

The four groups were as follows:

Model 1: CeramillZi, RoC (GE) – 0.45 mm (6 samples)

Model 2: CeramillZi, RoC (GE) – 0.90 mm (6 samples)

Model 3: Lava Plus, RoC (GE) – 0.45 mm (6 samples)

Model 4: Lava Plus, RoC (GE) – 0.90 mm (6 samples)

III A. FABRICATION OF CERAMILL ZI CAD/CAM COPINGS

i. Designing of the FPD copings

A single metal die was used for scanning. The metal die was sprayed with the Ivoclar contrast spray. The die was then placed in the Ceramill Map 300 scanner (Fig.17) for scanning. First 2D image was formed (Fig. 18) which was then converted to 3D image by the software (Fig. 19). The software used for designing was Ceramill Mind Software. The 3D image was used for designing and fabricating the 3 unit FPD copings (Fig. 20 and 21a&b). The designing was done by the lab professional. The designed image was saved in STL (StereoLithography) format.

ii. Milling of the FPD copings

On satisfactory designing, the FPD copings were milled. Two CeramillZi D shaped blanks were used (Fig. 22). One for 0.45 mm RoC at GE and another for 0.9 mm RoC at GE. Each blank yielded 6 samples of 3 unit FPD copings. A new blank was placed in the Ceramill Motion 2 (4X) machine at the start of the milling. The time required to mill one sample of 3 unit FPD coping was 30 minutes. (Fig. 23)

iii. Sintering of the FPD copings

After the 12 samples of 3 unit FPD coping were milled, 6 samples with RoC 0.45 mm (Fig. 24) and 6 samples with RoC 0.9 mm (Fig. 25) were placed in CeramillTherm machine for 8 hours at a temperature of 1450^{o}C for sintering (Fig. 26) (according to the manufacturer’s recommendation)

III B FABRICATION OF LAVA PLUS CAD/CAM COPINGS

i. Designing of the FPD copings

The already designed FPD coping saved in STL format was used. The STL file was compatible with the Exocad software used for designing the Lava plus coping.

ii. Milling of the FPD copings

On satisfactory designing, the FPD copings were milled. Two Lava Plus rectangular shaped blanks were used (Fig.27). One for 0.45 mm RoC at GE and another for 0.9 mm RoC at GE. Each blank yielded 6 samples of 3 unit FPD copings. A new blank was placed in the LAVA CNC 240 milling unit machine at the start of the milling. The time required to mill one sample of 3 unit FPD coping was 25 minutes. (Fig. 28)

iii. Sintering of the FPD copings

After the 12 samples of 3 unit FPD copings were milled, 6 samples with RoC 0.45 mm (Fig. 29) and 6 samples with RoC 0.9 mm (Fig. 30) were placed in sintering machine for 8 hours at a temperature of 1450^{o}C for sintering (according to the manufacturer’s recommendation)

IV CEMENTATION OF THE SAMPLES

After the FPD copings were fabricated, they were seated on the metal dies to check for the fit. The 24 bridge copings were then cemented to 24 metal dies using resin modified GIC cement. The cement was manipulated according to the manufacturer’s recommendation. The cement was applied to the internal surface of the coping and slowly seated over the die (Fig. 31). Cement was allowed to set for 60 seconds under a constant load of 15 N (Fig. 32).[4]The excess flash was removed with sharp explorer. The cement was allowed to set for 24 hours.

V. TESTING SAMPLES UNDER UNIVERSAL TESTING MACHINE

The static load bearing test was carried out using Universal testing machine to assess the fracture resistance of the copings at (METMECH Engineers, Chennai). The load was applied on each sample at the center of the pontic at a speed of 1 mm/min until the fracture of specimen occurred (Fig. 33). Load at fracture was automatically recorded by the software of the machine and the fracture curve was generated. For each curve, the first drop was marked, and the corresponding load was recorded as load at fracture.

VI. STATISTICAL ANALYSES

All the values were recorded as N/mm^2.

Statistical analysis were performed in Statistical Package for Social Sciences software (SPSS version 17, USA) using a personal computer. Data comparison was done by applying specific statistical tests to find out the statistical significance of the obtained results. The normality of the data was checked using Kolmogrov Smirnoff test and Levine’s test. Based on the normality, data between two groups and within [CeramillZi and Lava Plus, 0.45 mm and 0.90 mm] were compared using Unpaired Student’s T test or Mann Whitney U Test.

FIGURES

Fig 1 -Stainless steel die

Fig 2 - Ceramill Zi Blank

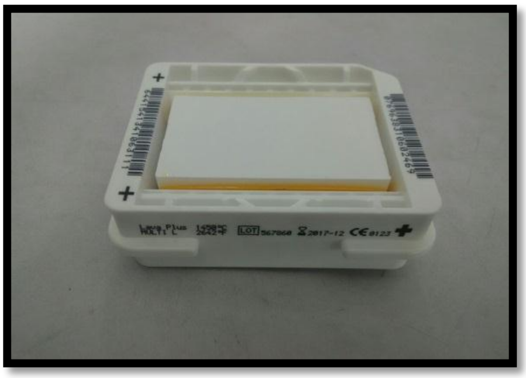


Fig 3 - Lava plus Blank

Fig 4 - Contrast Spray

Fig 5 - Resin cement

Fig 6 – Makino Vertical machine

Fig 7 - Ceramill map scanner

Fig 8 - Ceramill motion milling

Fig 9 - Ceramill therm sintering unit

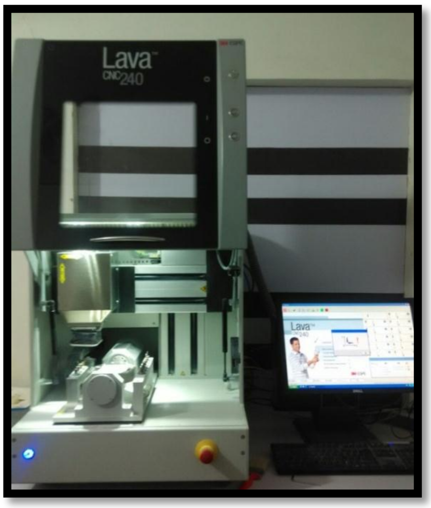


Fig 10 - Lava CNC 240 milling

Fig 11 - Nabertherm sintering unit

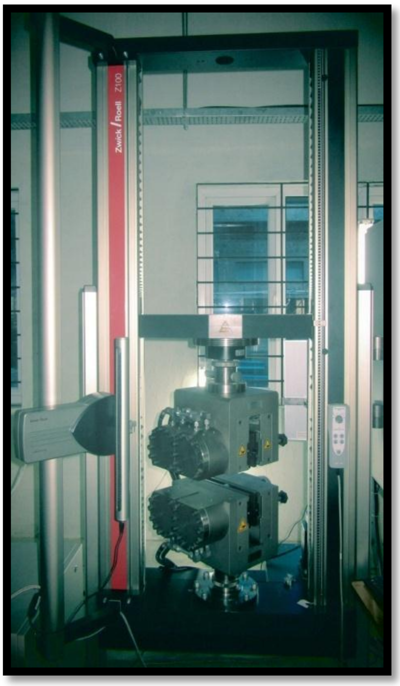


Fig 12 - Universal testing machine

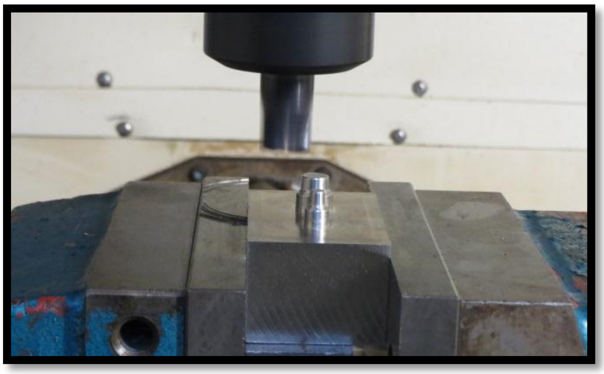


Fig 14 – Fabrication of stainless steel metal die

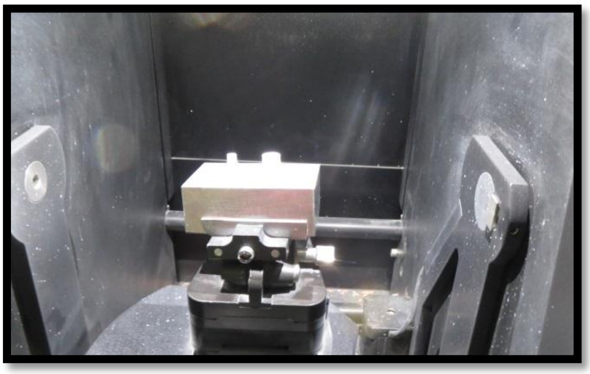


Fig 17 – Scanning of single metal die in Ceramill scanner

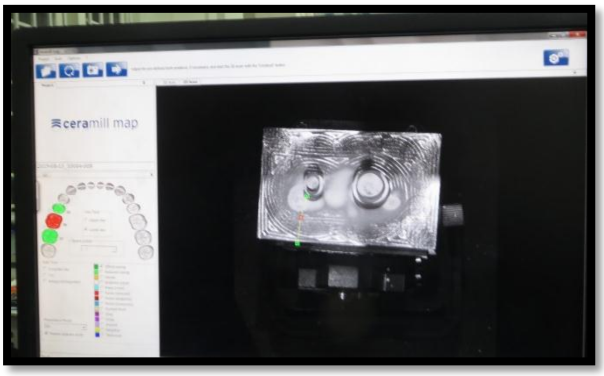


Fig 18 – 2D Image of the die

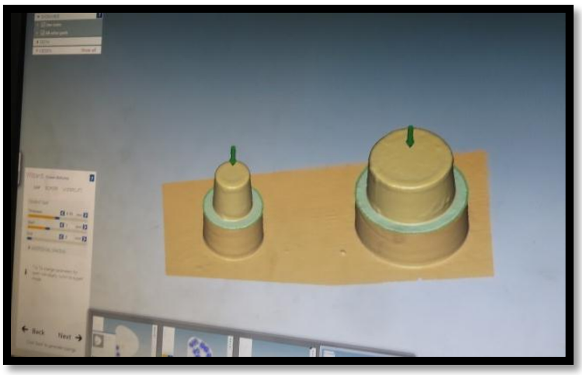


Fig 19 – 3D Image of the die

Fig 20 – CAD designing of 3 unit coping

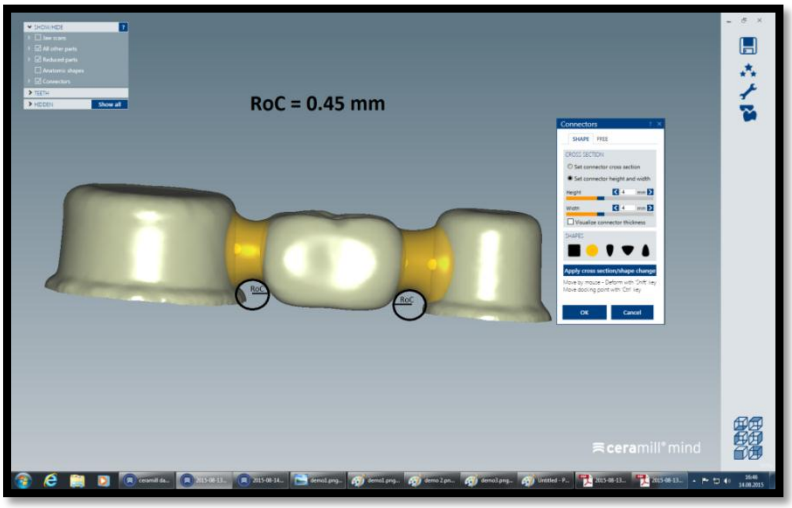


Fig 21.a – CAD designing of RoC 0.45mm

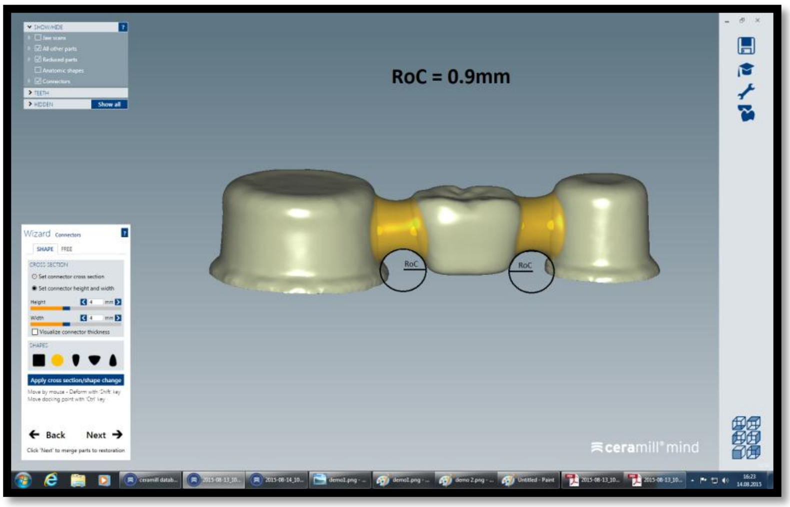


Fig 21.b – CAD designing of RoC 0.9 mm

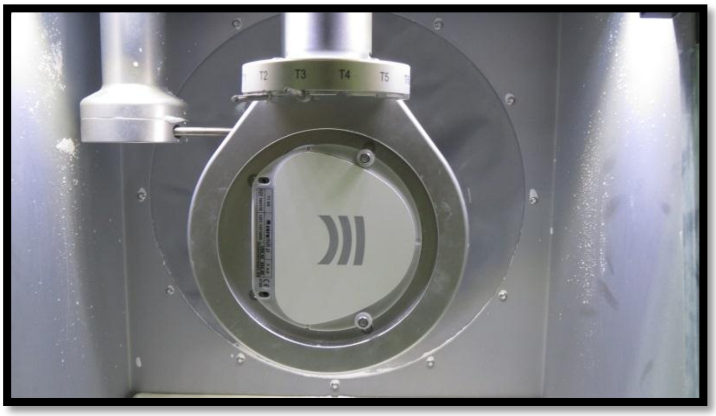
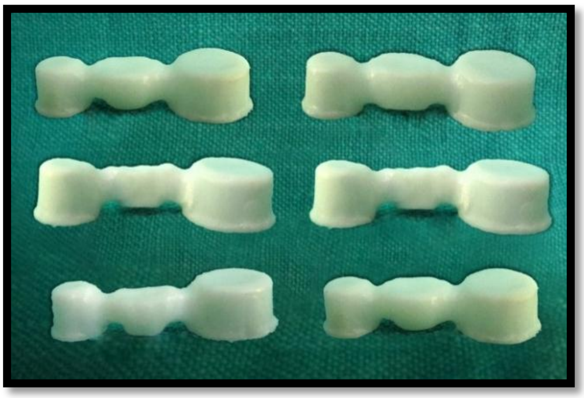


Fig 22 – Ceramill D shape blank placed in milling machine

Fig 23 – Milling of single ceramill3 unit coping



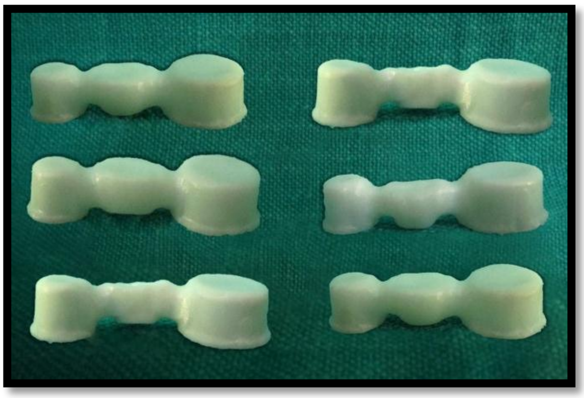


Fig 24 – Samples of Ceramill 3 unit copings with 0.45 mm RoC

Fig 25 – Samples of Ceramill 3 unit copings with 0.9 mm RoC

Fig 26 – Heat treatment of coping in sintering machine

Fig 27 – Lava plus rectangular shape blank placed in milling machine

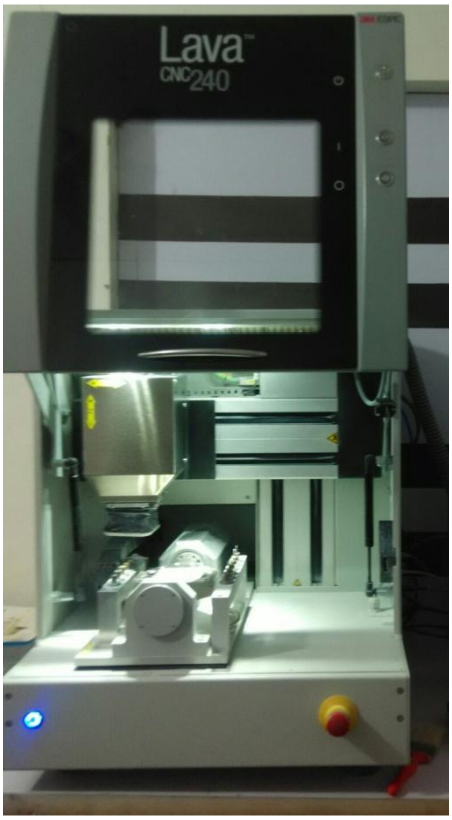


Fig 28 – Milling of Lava plus blank

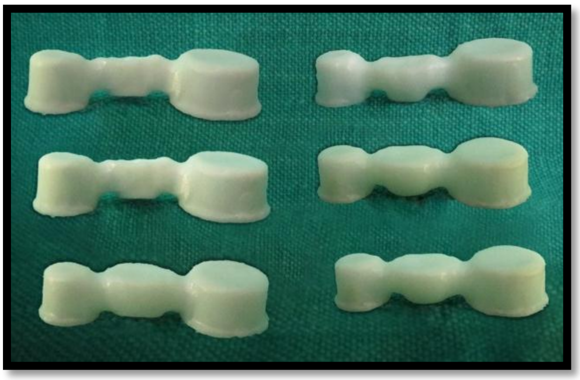


Fig 29 – Samples of Lava 3 unit copings with 0.45 mm RoC

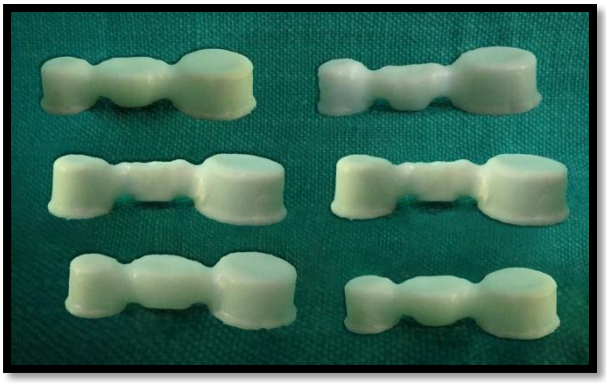


Fig 30 – Samples of Lava 3 unit copings with 0.9 mm RoC

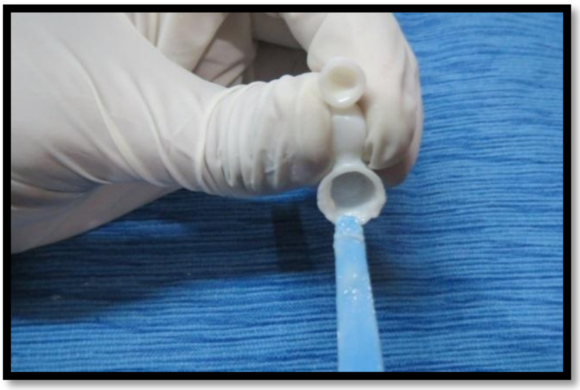


Fig 31 – Cement applied to the inner surface of coping

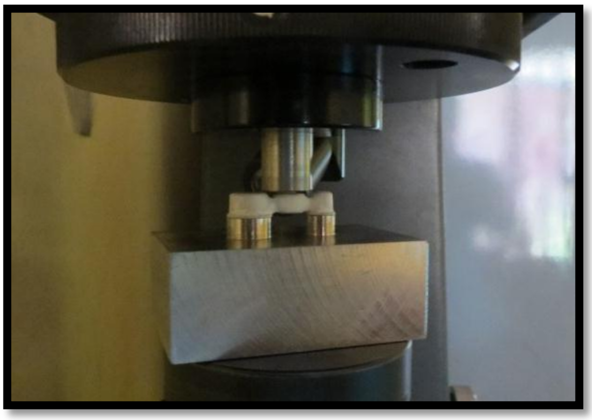


Fig 32 – Cementation under constant load of 15 N

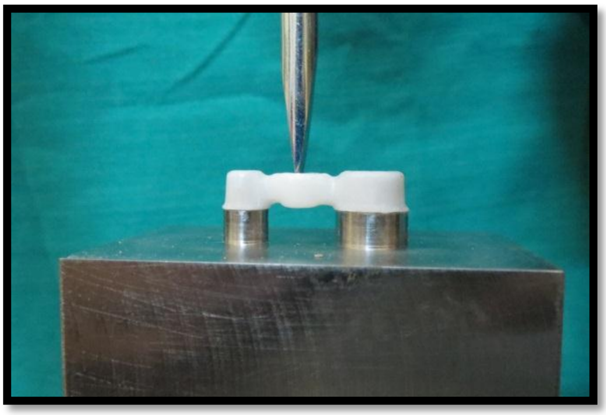


Fig 33 – Load applied at pontic in universal testing machine

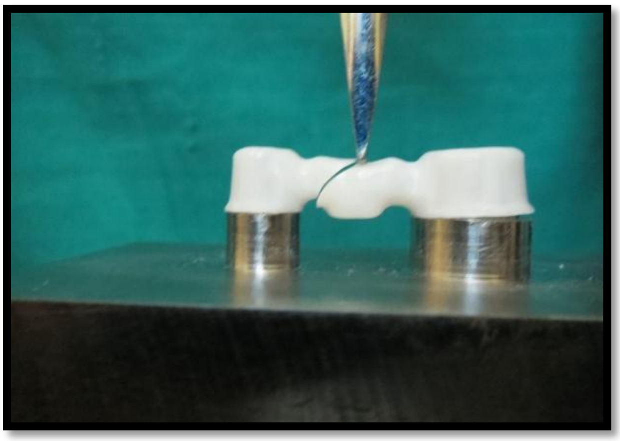


Fig. 35 – Oblique orientation of fracture path extending from gingival embrasure (GE) to occlusal contact area.

Note that occlusal embrasure is not included in fracture path.

RESULTS

The objectives of the present study were:

1. To determine the fracture resistance of CeramillZi three unit All ceramic bridge connector in 0.45 mm radius of curvature at the gingival embrasure.
2. To determine the fracture resistance of Lava Plus three unit All ceramic bridge connector in 0.45 mm radius of curvature at the gingival embrasure.
3. To determine the fracture resistance of CeramillZi three unit All ceramic bridge connector in 0. 9 mm radius of curvature at the gingival embrasure.
4. To determine the fracture resistance of Lava Plus three unit All ceramic bridge connector in 0.9 mm radius of curvature at the gingival embrasure.
5. To compare the fracture resistance of CeramillZi and Lava Plus three unit All ceramic bridge connector in 0.45 mm and 0.9 mm radii of curvature at the gingival embrasure.

The null hypothesis tested was
"the connector design with differing radii of curvature (0.45 mm and 0.90 mm) will not have an effect on the fracture resistance of the 3 unit FPD framework copings of both the materials (CeramillZi and Lava Plus). "

The data obtained from this study was subjected to Kolmogrov Smirnoff test to assess the normality. The test revealed that the data was not normally distributed. Hence, non parametric test was used. Mann Whitney U test was used as test of significance since, there were two independent groups for each comparison. If p value was < 0.05, it was considered significant (there is a statistical difference between the groups). If not, the groups were considered similar to each other (there was no difference between the groups).

The mean and standard deviation of each group was assessed using descriptive statistics. All values are recorded in term of Newton (N)

<u>Descriptive Tables</u>

Table 1: Compressive load (N) for CeramillZi, 0.45 mm RoC

S. No	Compressive load (N)	Minimum	Maximum	Mean	SD
1	875	875	1208	1062.17	123.80
2	1183				
3	1208				
4	987				
5	1054				
6	1066				

SD - Standard deviation

For Group 1 (CeramillZi, 0.45 mm RoC) the mean compressive load was 1062.17 ± 123.80 N with a minimum of 875 N and maximum of 1208 N. Graph 1 represents the compressive load of six samples in the group.

Graph 1

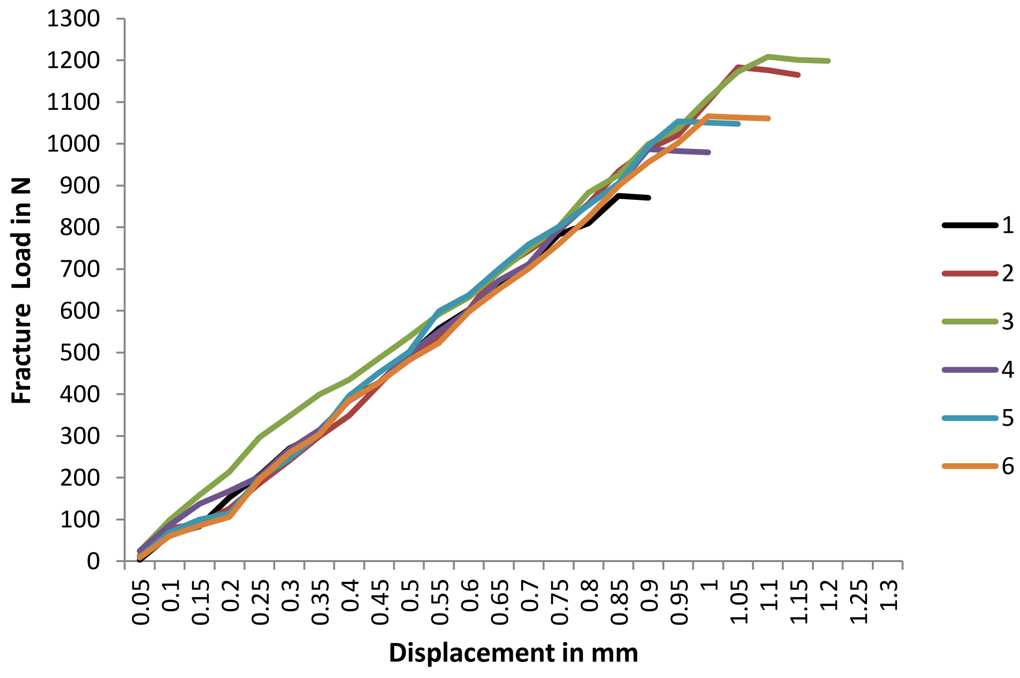


Table 2: Compressive load (N) for CeramillZi, 0.90 mm RoC

S. No	Compressive load (N)	Minimum	Maximum	Mean	SD
1	1234	1109	1320	1243.00	72.35
2	1250				
3	1109				
4	1320				
5	1258				
6	1287				

SD - Standard deviation

For Group 2 (CeramillZi, 0.9 mm RoC) the mean compressive load was 1243.00 ± 72.35 N with a minimum of 1109 N and maximum of 1320 N. Graph 2 represents the compressive load of six samples in the group.

Graph 2

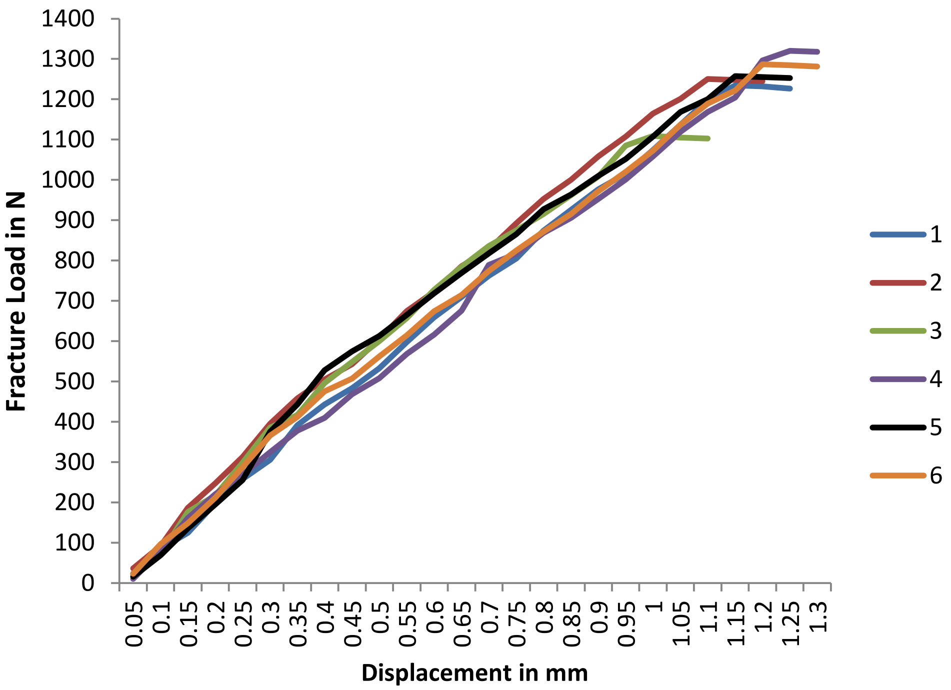


Table 3: Compressive load (N) for Lava Plus, 0.45 mm RoC

S. No	Compressive load (N)	Minimum	Maximum	Mean	SD
1	1100	950	1100	1039.00	62.33
2	1098				
3	950				
4	986				
5	1025				
6	1075				

SD - Standard deviation

For Group 3 (Lava Plus, 0.45 mm RoC) the mean compressive load was 1039.00 ± 62.33 N with a minimum of 950 N and maximum of 1100 N. Graph 3 represents the compressive load of six samples in the group.

Graph 3

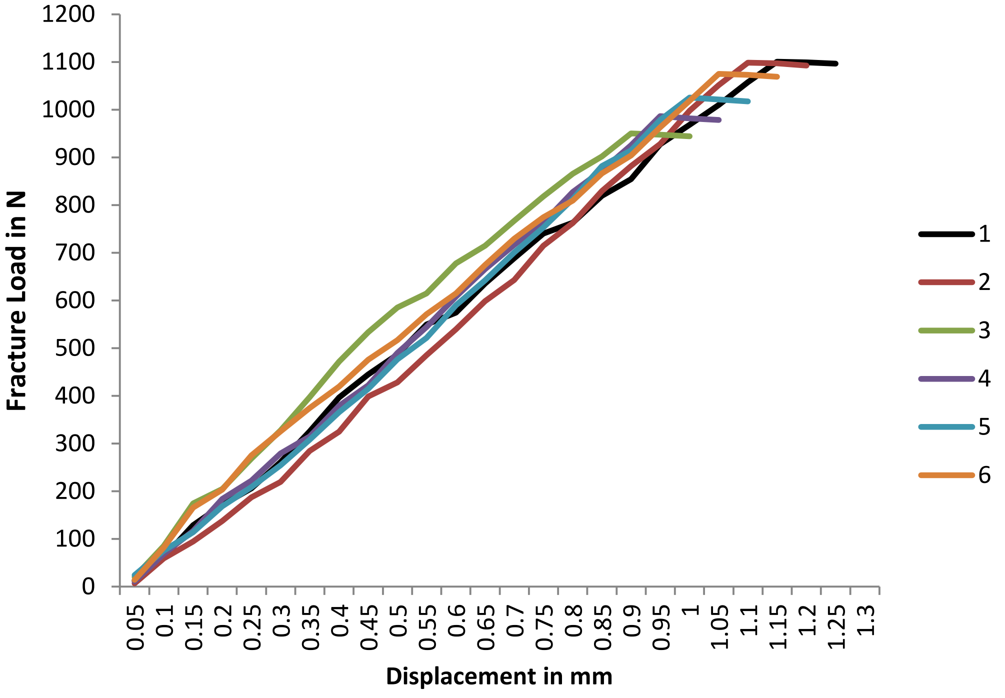


Table 4: Compressive load (N) for Lava Plus, 0.90 mm RoC

S. No	Compressive load (N)	Minimum	Maximum	Mean	SD
1	1156	1125	1300	1237.50	76.50
2	1125				
3	1269				
4	1282				
5	1300				
6	1293				

SD - Standard deviation

For Group 4 (Lava Plus, 0.90 mm RoC) the mean compressive load was 1237.50 ± 76.50 N with a minimum of 1125 N and maximum of 1300 N. Graph 4 represents the compressive load of six samples in the group.

Graph 4

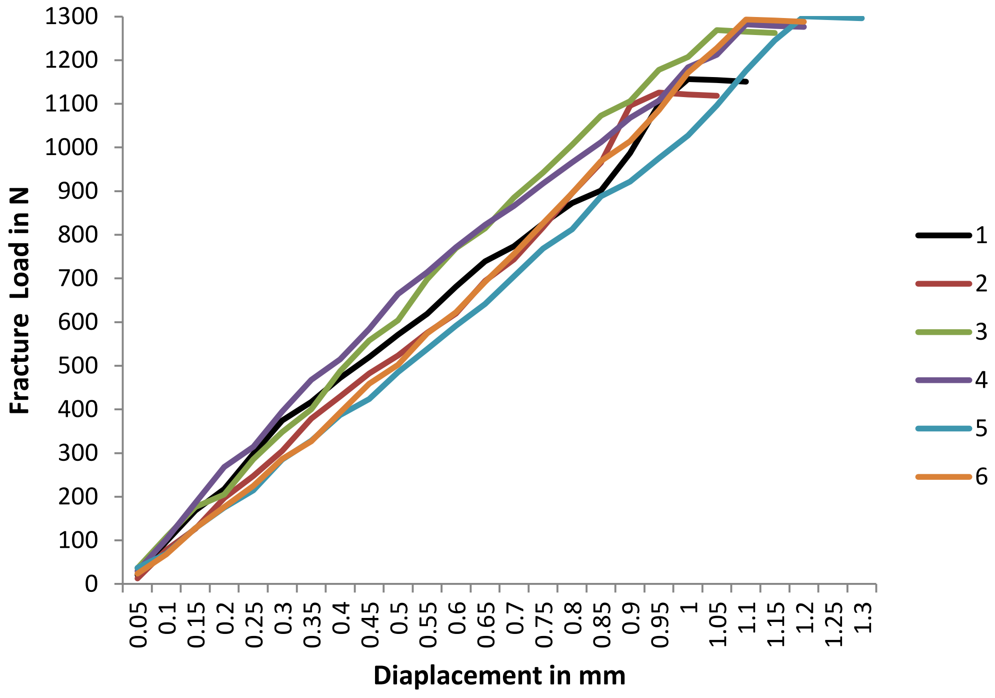
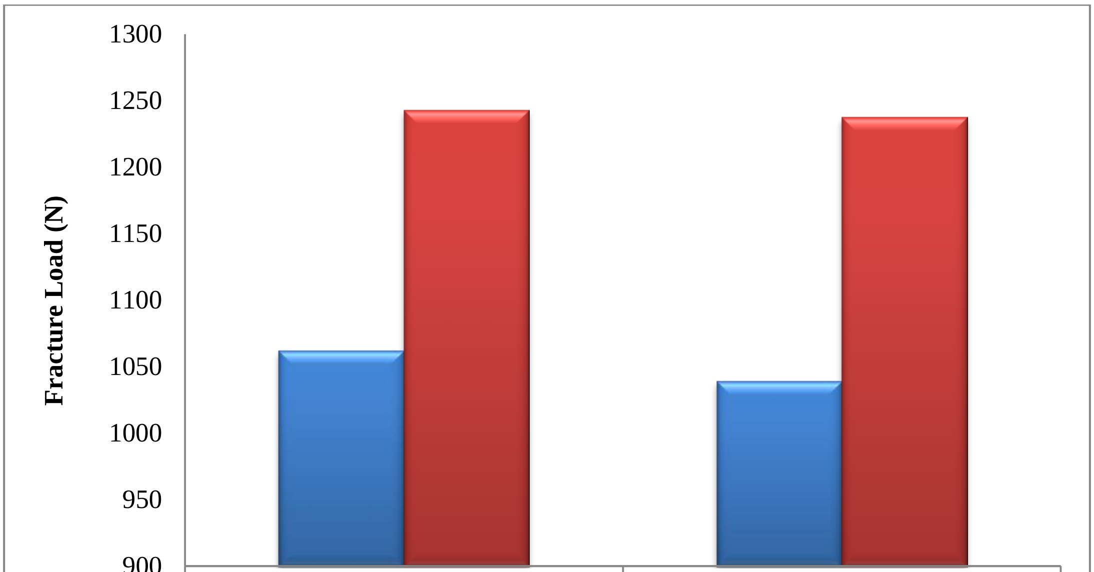


Table 5: Number of specimens fractured at mesial and distal areas of connector

Designs		Location of breakage	
Group	No. of samples	Mesial (%)	Distal (%)
I	6	4 (80%)	2 (20%)
II	6	3 (50%)	3 (50%)
III	6	2 (20%)	4 (80%)
IV	6	3 (50%)	3 (50%)

Graph 5: Comparative bar graph showing the Mean Compressive Load (N) for 0.45 mm and 0.90 mm RoC for CeramillZi and Lava Plus



	Ceramill Zi	Lava Plus
0.45 mm	1062.17	1039
0.90 mm	1243	1237.5

<u>Inferential Tables</u>

Table 6: Comparison of 0.45 mm and 0.90 mm RoC CeramillZi using Mann Whitney U test

RoC	Mean ± SD	U Statistic	Z statistic	p value
0.45 mm	1062.17 ± 123.80	2.00	-2.56	**0.010 ***
0.90 mm	1243.00 ± 72.35			

SD - Standard deviation
U, Z statistic and p value obtained from Mann Whitney U test
* p value < 0.05 is significant

On comparing the CeramillZi 0.45 mm and 0.90 mm RoC ,there was a statistically significant difference between the 0.45 mm and 0.90 mm groups at p=0.010.

Inference:

The CeramillZi groups are dissimilar. The 0.90 mm RoC group had a higher fracture resistance when compared to 0.45 mm RoC.

Table 7: Comparison of 0.45 mm and 0.90 mm RoC Lava Plus using Mann Whitney U test

RoC	Mean ± SD	U Statistic	Z statistic	p value
0.45 mm	1039.00 ± 62.33	0.00	-2.882	**0.004 ***
0.90 mm	1237.50 ± 76.50			

SD - Standard deviation
U, Z statistic and p value obtained from Mann Whitney U test
* p value < 0.05 is significant

On comparing the Lava Plus 0.45 mm and 0.90 mm RoC ,there was a statistically significant difference between the 0.45 mm and 0.90 mm groups at p=0.004.

Inference:

The Lava Plus groups are dissimilar. The 0.90 mm RoC group had a higher fracture resistance when compared to 0.45 mm RoC.

Table 8: Comparison of CeramillZi and Lava Plus for 0.45 mm RoC using Mann Whitney U test

Material	Mean ± SD	U Statistic	Z statistic	p value
CeramillZi	1062.17 ± 123.80	15.78	-.320	.748
Lava Plus	1039.00 ± 62.33			

SD - Standard deviation
U, Z statistic and p value obtained from Mann Whitney U test
* p value < 0.05 is significant

On comparing the 0.45 mm RoC between CeramillZi and Lava Plus, there was statistically no significant difference between both the materials with p value > 0.05.

Inference:

The groups are similar to each other. The difference between the groups is less and also not statistically significant. Therefore, both the companies (CeramillZi and Lavaplus) have similar fracture resistance for 0.45 mm RoC

Table 9: Comparison of CeramillZi and Lava Plus for 0.90 mm RoC using Mann Whitney U test

Material	Mean ± SD	U Statistic	Z statistic	p value
CeramillZi	1243.00 ± 72.35	16.00	-.320	.748
Lava Plus	1237.50 ± 76.50			

SD - Standard deviation
U, Z statistic and p value obtained from Mann Whitney U test
* p value < 0.05 is significant

On comparing the 0.90 mm RoC between CeramillZi and Lava Plus, there was statistically no significant difference between both the materials with p value > 0.05.

Inference:

The groups are similar to each other. The difference between the groups is less and also not statistically significant. Therefore, both the companies (CeramillZi and Lavaplus) have similar fracture resistance for 0.90 mm RoC.

DISCUSSION

DISCUSSION:

Contemporary world is more time and beauty conscious. This attitude has also crept into prosthetic dentistry. With the technological advancements in the dental science, it is now possible to achieve this particular demand by using all ceramics.

The most commonly used all ceramic material for posterior CAD CAM restorations is yttrium-stabilized tetragonal zirconia polycrystals (Y- TZP), because of its higher strength, and higher toughness than other dental ceramics. It is biocompatible and fulfills the aesthetic demand. Flexural strength and fracture toughness of yttrium stabilized zirconia is 900 to 1200 MPa and 9 – 10 $Mpa/m^{1/2}$respectively[42]. Y-TZP cores are glass free, and because they have a polycrystalline microstructure they do not exhibit stress corrosion[43]. However, despite its favourable features, it cannot be manipulated by usual techniques like casting, sintering, hot pressing or grinding. A way to overcome this limitation is CAD CAM process. This process is more economical due to reduction in fabrication steps, fabrication errors and overall labour hours required.

While designing a three unit All ceramic CAD-CAM restoration with Y-TZP, proper designing is required to reduce stress concentration as ceramics are highly susceptible to tensile stress.

The most common mode of failure of All ceramic FPD is the fracture of connectors, as it is the thinnest section of a framework and will bend more easily than the pontic and retainers.[16]In-vitro tests and finite element analysis (FEA) have also shown that the highest level of stress concentration occur in the connector area of All-ceramic Fixed Partial Dentures (FPDs) [3,16,35]

A connector is defined in fixed dental prosthodontics, as the portion of a fixed partial denture (FPD) that unites the retainer(s) and pontic(s)[58]. When occlusal forces are applied directly through the long axis of a three unit All-ceramic bridge, compressive stresses develop at the occlusal aspect of the connector at the marginal ridge, and tensile stresses develop at the gingival surface of the connector[36]. As ceramics are weak under tensile stresses, this contributes to the propagation of microcracks located at the gingival surface of the connector through the core material in an occlusal direction leading to fracture[39]. Therefore, the dimensions of FPD connectors must be large enough to counteract the concentrations of stress that develop in the framework.

While designing a connector for 3 unit All ceramic CAD CAM restoration, various features such as connector height, width, cross sectional design and area, radius of curvature at occlusal and gingival embrasure, have to be taken into consideration. Clinically, the connector design should be determined according to material properties, anatomical limitations, hygiene considerations and esthetic expectations.

In order to improve the survival of FPD restoration, it is desirable to make the cross sectional area of the framework connector as large as possible, regardless of the material. The 3 x 3 mm and 4 x 4 mm connector dimensions were proposed for All ceramic systems.

With regards to the cross sectional design of the connector, Singh MS, Tuli AK[54] stated that the maximum stress concentration was seen with circular cross sectional design while the least was with inverted T shape connector. Manufacturer's recommend that the connector cross sectional area should be over 9 mm^2 for Y-TZP crowns on molars.

The radius of curvature at a given point is the radius of a circle that mathematically best fits the curve at that point (Fig. 35). In order to reduce the fracture probability when designing All-ceramic FPDs, the shape of the connector is an important factor to consider. In particular, the radius of curvature at the gingival embrasure plays a significant role in the load-bearing capacity as tensile stresses concentrate here during loading.

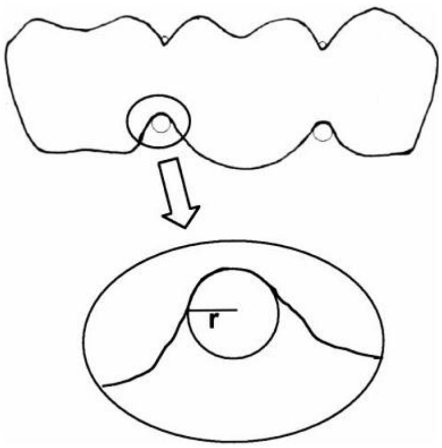


Fig. 35 - Schematic diagram of Radius of Curvature (RoC)

FPDs with small gingival embrasure radii are subjected to high stress concentrations in the connector area which results in lower fracture resistance compared to FPDs with larger embrasure radii. [4,10,31,56]

Oh et al[35] demonstrated in a finite element and a fractographic analyses that connector fracture was initiated at the gingival embrasure and that a larger radius of curvature at the gingival embrasure reduces the concentration of tensile stresses, thus increasing the fracture resistance of the hot pressed core ceramic FPD. To reduce the stress concentration at the gingival embrasure, they have proposed that gingival embrasure curvature radius be of 0.45 mm. Bahat et al[4] has investigated the fracture strength of CAD/CAM 3 unit All- ceramic **anterior FPDs** made of Y- TZP based on different radii of curvature in the gingival embrasure.

The present study was conducted to know the fracture resistance of CAD / CAM designed Y-TZP **posterior** three unit All- ceramic bridge connectors with different radii of curvature in the gingival embrasure.

Two commercially available (CeramillZi and Lava plus) All ceramic three unit fixed partial denture bridges system were evaluated with two different radii of curvature (Roc 0.45mm and 0.90 mm at gingival embrasure). In the present study the veneer material was not used as it is difficult to achieve uniform build up in a standardized way for all the crowns. Another reason is that the veneering itself would not affect the fracture strength, since the elastic modulus of the veneering material is much lower than that of the core material.[4]

In this particular study, the connector dimensions were standardized and the radius of curvature at the gingival embrasure was varied. A connector dimension of 4 x4 mm with circular cross section[24] and height/width ratio of 1:1 was used. 24 three unit All ceramic bridge copings with the above specified dimension were fabricated and they were subjected to heat treatment in sintering machine to simulate veneering.[4] They were divided into four groups. Each group consisted of 6 samples. The groups were as follows:

Group 1:	CeramillZi, RoC	–	0.45 mm
Group II:	CeramillZi, RoC	–	0.90 mm
Group III:	Lava Plus, RoC	–	0.45 mm
Group IV:	Lava Plus, RoC	–	0.90 mm

The All ceramic 3 unit bridges were cemented on their respective metal dies using resin modified GIC based on previous study conducted by Ernst et al.[16]The resin modified GIC was mixed according to the manufacturer's instruction and applied on to the inner surface of bridge copings. It was then seated on the metal die and the cement was allowed to set for 60 seconds under a constant load of 15 N.[4]

For determination of fracture resistance under loading the bridge specimens were loaded axially to fracture in a universal testing machine. The load was applied at the center of the pontic at a speed of 1.0 mm/min with a flat end stainless steel rod to have invariable contact surface, that guarantee a steady increase of load during the test[6]. Loading was continued to the point of fracture and failure loads were recorded with computer software. The crack initiation point on the load versus speed curve chart for All ceramic FPDs was determined by a sharp decrease in the loading curve and confirmed by an audible sound. The fracture location occurred equally in the mesial and distal connector of the pontic in all the groups. (Table-5).The propagation of crack pathways were oblique from gingival embrasure to occlusal direction through the connector and pontic without involving the occlusal embrasure (Fig. 34). It is very similar to the fracture pattern reported for posterior ceramic FPDs in literature[36]. These findings indicate that the curvature at the occlusal embrasure has no effect on fracture resistance, validating that the occlusal embrasure (OE) can be made sharp to improve esthetics of posterior FPDs.

In the present study, the mean fracture resistance of the various groups were as follows. Group 1 (CeramillZi, RoC 0.45 mm) – 1062 ± 123.80N, Group 2 (CeramillZi, RoC0.90 mm) – 1243 ± 72.35N, Group 3 (Lava Plus, RoC 0.45 mm) – 1039 ± 62.33N, Group 4 (Lava Plus, RoC 0.90 mm) – 1237 ± 76.5N. Groups 2 and 4, which had a larger curvature at the GE (RoC = 0.90 mm) exhibited significantly higher fracture resistance than groups 1 and 3 which had a smaller curvature at the GE (RoC = 0.45 mm). Studies report that biting force in first molar area is 400-890N with force exceeding 1000N during parafunction[38]. The forces recorded in this study by all the groups are comparatively higher than the normal biting forces.

The results obtained were statistically analyzed. In the present study, non parametric Mann Whitney U test was used to compare two groups. Mann Whitney U test was used since the data distribution was not normal, the sample size was small and was not randomly selected. A p value less than or equal to 0.05 was considered significant.

In the present study, there was a statistically significant difference between the Groups 1 and 2 and Groups 3 and 4. The fracture resistance increased by 17.02% and 19.06% in Groups 2 and 4 respectively. On a whole, there was 18.05% increase in fracture resistance when the radius of curvature was changed to 0.90 mm from 0.45 mm. Hence, the null hypothesis stating that the connector design with differing radii of curvature (0.45 mm and 0.90 mm) will not have an effect on the fracture resistance of the 3 unit FPD copings of both the materials (CeramillZi and Lava Plus) was rejected. The increase in fracture resistance of the 3 unit All ceramic bridges with increase in radius of curvature is in accordance with studies conducted by Oh and Anusavice[36], Oh, Gotzen and Anusavice[35], Naseef, Khalil and Khader[56]and Bahat, Mahmood and Steyern.[4]

Limitations of the study:

1. The sample size was not scientifically determined.
2. Factors such as effect of humidity, temperature, occlusal stress etc., were not taken into consideration in this study.
3. Effect of finish lines was not taken into consideration.
4. Various other vector forces like lateral forces, oblique forces were not taken into account.
5. Concept of cyclic loading was also not considered.
6. The elastic modulus of the supporting structure is not taken into account because all samples were cemented on a metal die, and the study evaluated only the mechanical property of two different companies.

Further studies may be conducted to overcome the limitations of the present study.

CONCLUSION

Within the limitations of this invitro study, the following conclusions were drawn:

1. By increasing the radius of curvature from 0.45 mm to 0.90 mm, the fracture resistance of 3 unit All ceramic bridge copings for CeramillZi increased by 17.02%.
2. By increasing the radius of curvature from 0.45 mm to 0.90 mm, the fracture resistance of 3 unit All ceramic bridge copings for Lava Plus increased by 19.06%
3. The results of this study suggest that the occlusal embrasure can be designed as sharp as is practical for the esthetics of an All ceramic FPD, provided that the gingival embrasure has a greater radius of curvature.
4. For short clinical crown where connector height is limited, radius of curvature at the gingival embrasure can be increased upto 0.9 mm, in order to increase fracture resistance.

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